Further, the receiver platform can communicate its unique address, code or other information to the Tx by data transmission.

[0058] Time multiplexing can be used by the Tx to power-up and communicate with the receiver platforms. In certain embodiments, the Tx can perform multi-lobe beam-forming to simultaneously power-up and/or communicate with multiple receiver platforms. The Tx could also use multiple frequencies to simultaneously power-up and/or communicate with multiple receiver platforms. In such embodiments, the link between each Tx and receiver platforms can be dynamically reconfigured for optimizing efficiency separately for each link.

[0059] Multiple Tx devices can be employed to power-up a single or multiple receiver platform devices to offer more flexibility for beam-forming, as well as to enable sufficient power transfer without exceeding any regulations. A Tx may be composed of a single element or a multi-element array.

[0060] In preferred embodiments, the receiver unit includes a back-side structure having low acoustic impedance (i.e., acoustic impedance of 10 Mrayl or less, more preferably 2 Mrayl or less. FIG. **11** shows an example, where the low acoustic impedance is provided by a back side air gap. Here "Acoustic impedance" is specific acoustic impedance having SI units (Pa s/m) also denoted as Rayl.

B) Experimental Demonstration

B1) Introduction

[0061] It has been proposed that implantable medical devices (IMDs) employing neuromodulation therapies, or "electroceuticals," may supplant drugs as the primary treatment for many neurological disorders. Unlike drugs which freely diffuse about the body, neuromodulation therapies are more targeted, allowing for the mitigation of unwanted side-effects. There are already many neuromodulation devices on the market or in development to treat disorders like Parkinson's and chronic pain; however, some of them are large, invasive, and prone to causing infection. In order to alleviate these issues, researchers are attempting to shrink the implants down to millimeter or sub-mm sizes and replace bulky batteries with reliable and highly efficient wireless power links. Most of these researchers have focused on RF or inductive powering, but as we have described in previous studies, ultrasonic power delivery has several key advantages over conventional RF and inductive powering when shrinking down to the mm-scale. Namely, ultrasound undergoes relatively small propagation losses through tissue (~1 dB·MHz/cm) and has a high FDA allowed time-averaged intensity (7.2 mW/mm^2), making it ideal for efficient power transmission at great depths (>5 cm). Additionally, ultrasound has small wavelengths in tissue (e.g. 1.5 mm at 1 MHz) allowing for superior energy focusing down to mm-spots, as well as more efficient energy recovery from a ultrasonic receiver.

[0062] Current and future IMDs may be equipped with several functionalities, such as electrical or optical stimulation, neural recording, and temperature and pressure sensing within one module—these functions require a large range of average implant load (P_{load}) typically ranging from 10 μW to 1 mW. In addition, next-generation IMDs will be programmable with duty-cycled operation and different functional modes, leading to dynamically varying P_{load} for an individual IMD. Static links can become inefficient with large load perturbations due primarily to impedance mismatch between the power receiver and the non-linear power recovery chain. As demonstrated below, an implant optimally matched for 1 mW achieves less than 5% efficiency when operated at 10 μW. Low efficiency is a major reliability problem, leading to significantly reduced battery life of the external source and potential loss of function of the IMD if the required power cannot be achieved. Therefore, an ideal power receiver should be tunable, along with the source, to maximize the power matching efficiency over a wide variety of applications and dynamic loads.

[0063] With proper choices of material and dimensions, a piezoelectric ultrasonic receiver can be designed to be mm-sized with an optimal electrical impedance for a highly variable load. In addition, by using frequency as a degree of freedom, we demonstrate off-resonance operation to modulate the receiver impedance for adaptive matching. In contrast, mm-sized implantable antennas, which are typically operated in the low-GHz range to combat tissue loss, offer much smaller radiation resistance and efficiency, due to mismatch in aperture and wavelengths as well as dielectric loading.

[0064] In this section, we first introduce the implant power recovery chain for an IMD, and describe the impedance match interface between the piezoelectric receiver and implant loads. We consider the effect of average P_{load} on the input impedance of the power recovery circuit and demonstrate the concept of off-resonance operation for adaptable impedance tuning. Then a design procedure is presented to achieve the impedance specifications with a piezoelectric receiver. The selection of material and dimensions for ultrasonic receivers greatly influences frequency of operation and the impedance tuning range, so several different materials, including bio-compatible options, are compared. Finally, we use two adaptive matching topology examples to show significant improvement in the total implant efficiency over a non-tunable power recovery chain.

B2) Power Recovery Chain Under a Variable Load

[0065] A schematic diagram of an ultrasonic power recovery chain for an IMD in the steady state is shown in FIG. **8**, which includes a piezoelectric receiver, a matching network, power recovery circuits, and an average application load. The total implant efficiency ($\eta_{implant}$) is determined by three major components: the acoustic-to-electrical power conversion efficiency of the receiver (PCE), the efficiency of the power recovery circuit ($\eta_{AC\text{-}DC}$) and the power matching efficiency (PME) between the first two components. Therefore, $\eta_{implant}$ can be represented as

$$\eta_{implant} = \frac{P_{load}}{P_{acou}} = PCE \cdot PME \cdot \eta_{AC-DC}, \tag{1}$$

where P_{acou} is the total incident acoustic power on top of the receiver. There is extensive literature on designing power electronics for power receivers to achieve high $\eta_{AC\text{-}DC}$; hence we focus on optimizing efficiency of the ultrasonic receiver and impedance matching interface due to the large variation of P_{load} in an IMD.

[0066] A first-order calculation can be made to model any non-linear power recovery circuits, along with the implant